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Quantitative determination of the femoral offset templating error in total hip arthroplasty using a new geometric model

E. F. Liechti,
M. C. Attinger,
A. Hecker,
K. Kuonen,
A. Michel,
F. M. Klenke

From Inselspital, Bern University Hospital, University of Bern, Bern, Switzerland

Aims

Traditionally, total hip arthroplasty (THA) templating has been performed on anteroposterior (AP) pelvis radiographs. Recently, additional AP hip radiographs have been recommended for accurate measurement of the femoral offset (FO). To verify this claim, this study aimed to establish quantitative data of the measurement error of the FO in relation to leg position and X-ray source position using a newly developed geometric model and clinical data.

Methods

We analyzed the FOs measured on AP hip and pelvis radiographs in a prospective consecutive series of 55 patients undergoing unilateral primary THA for hip osteoarthritis. To determine sample size, a power analysis was performed. Patients' position and X-ray beam setting followed a standardized protocol to achieve reproducible projections. All images were calibrated with the KingMark calibration system. In addition, a geometric model was created to evaluate both the effects of leg position (rotation and abduction/adduction) and the effects of X-ray source position on FO measurement.

Results

The mean FOs measured on AP hip and pelvis radiographs were 38.0 mm (SD 6.4) and 36.6 mm (SD 6.3) ($p < 0.001$), respectively. Radiological view had a smaller effect on FO measurement than inaccurate leg positioning. The model showed a non-linear relationship between projected FO and femoral neck orientation; at 30° external neck rotation (with reference to the detector plane), a true FO of 40 mm was underestimated by up to 20% (7.8 mm). With a neutral to mild external neck rotation (≤ 15°), the underestimation was less than 7% (2.7 mm). The effect of abduction and adduction was negligible.

Conclusion

For routine THA templating, an AP pelvis radiograph remains the gold standard. Only patients with femoral neck malrotation > 15° on the AP pelvis view, e.g. due to external rotation contracture, should receive further imaging. Options include an additional AP hip view with elevation of the entire affected hip to align the femoral neck more parallel to the detector, or a CT scan in more severe cases.

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Keywords: Total hip arthroplasty, AP pelvis radiograph, Templating, Geometric model, Thickness of the lesser trochanter, AP hip radiograph, Femoral offset

Correspondence should be sent to Emanuel F. Liechti; email: emanuel.liechti@insel.ch

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Introduction

The restoration of the physiological biomechanics of the hip joint in total hip arthroplasty (THA) is crucial. With the increasing options of modularity of the prosthetic components, the accurate adjustment of leg length and femoral offset (FO) has become feasible. These parameters have significant impact on the clinical outcome after THA. FO has been shown to correlate with abductor strength and range of motion.[1-5] Reduced FO leads to an inferior functional

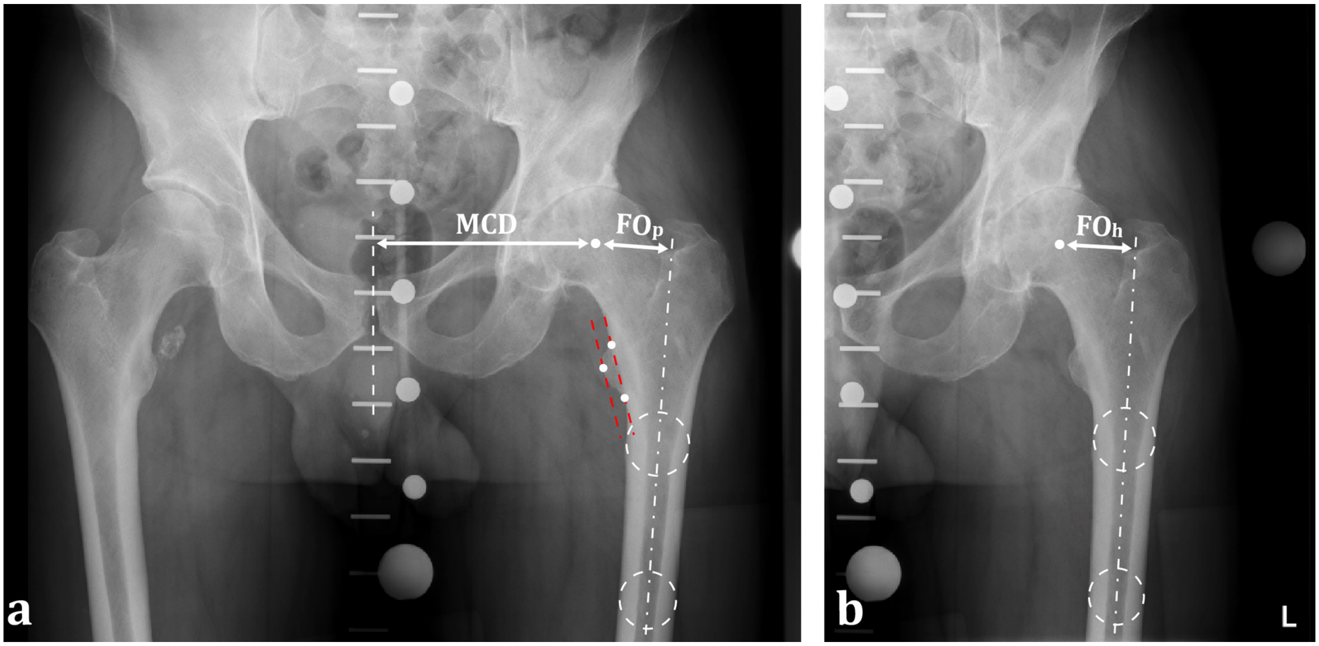

Fig. 1

a) Preoperative anteroposterior (AP) pelvis radiograph with KingMark calibration markers. Femoral offset (FO_p) was calculated as the perpendicular distance between the femoral shaft axis and the femoral head centre. The mid-centre distance (MCD) was defined as the shortest distance between the midline of the pelvis and the centre of the femoral head. Rotation of the femoral neck to the detector plane was assessed using the thickness of the lesser trochanter (TLT) defined by the perpendicular distance between the two dotted parallel lines; the proximal and distal cortical intersection between the lesser trochanter and the femoral cortex defined the first line, and the outer prominent contour of the lesser trochanter defined the second line. b) Preoperative AP hip radiograph with KingMark calibration markers and femoral offset measurement (FO_h).

outcome. Additionally, joint stability[2] and polyethylene wear,[6,7] as well as functional outcomes,[8] are adversely affected by reduced FO.

In clinical practice, the assessment of the FO relies on preoperative radiographs. Traditionally, low-centred anteroposterior (AP) radiographs of the pelvis have been performed to template THA. Several studies have shown that these standard radiographs might underestimate the FO when compared to CT scans, as the projection of the anteverted femoral neck underestimates the true offset.[9-11] In clinical practice, CT scan templating in THA is not feasible on a routine basis. Therefore, some authors have recommended additional AP radiographs of the hip to restore the FO more accurately. Merle et al[10] have shown that AP hip radiographs might reduce the projection-related underestimation of the FO. However, additional X-ray imaging increases the radiation exposure of the patient, as AP pelvis templating is still required to restore the leg length and the centre of rotation.

We hypothesized that AP hip radiographs are not required for routine THA templating in order to restore the native FO. The study therefore aimed to analyze the FO obtained by AP radiographs of the pelvis and the hip. Additionally, a geometric model was created to simulate the FOs measured on AP hip and AP pelvis radiographs, taking into account different leg positions. With this model, we aimed to establish quantitative data of the measurement error of the FO in relation to leg positioning and X-ray source position.

Methods

Radiological analysis. A consecutive series of preoperative radiographs of 55 patients undergoing unilateral primary THA for osteoarthritis (OA) of the hip (primary OA n = 46, secondary OA n = 9) was analyzed (prospective Level IV study). The study period was six months. All types of OA were included and documented. Patients undergoing bilateral THA and hips with deformity of the pelvis and/or the proximal femur due to Perthes' disease, hip dysplasia with dislocation of the centre of rotation, and post-traumatic deformity were excluded. Overall, 31 patients were male and 24 were female. The mean age was 60 years (23 to 88). The mean BMI was 27 kg/m^2 (18 to 59). All patients received an uncemented monoblock pressfit cup (RM Pressfit Cup, Mathys, Switzerland) and an uncemented (Fitmore Stem, Zimmer, Switzerland; n = 36) or a cemented (Centris Stem, Mathys; n = 19) stem. The study was approved by the regional ethical board committee.

Each patient obtained an AP radiograph of the pelvis and an AP radiograph of the affected hip. Patients were

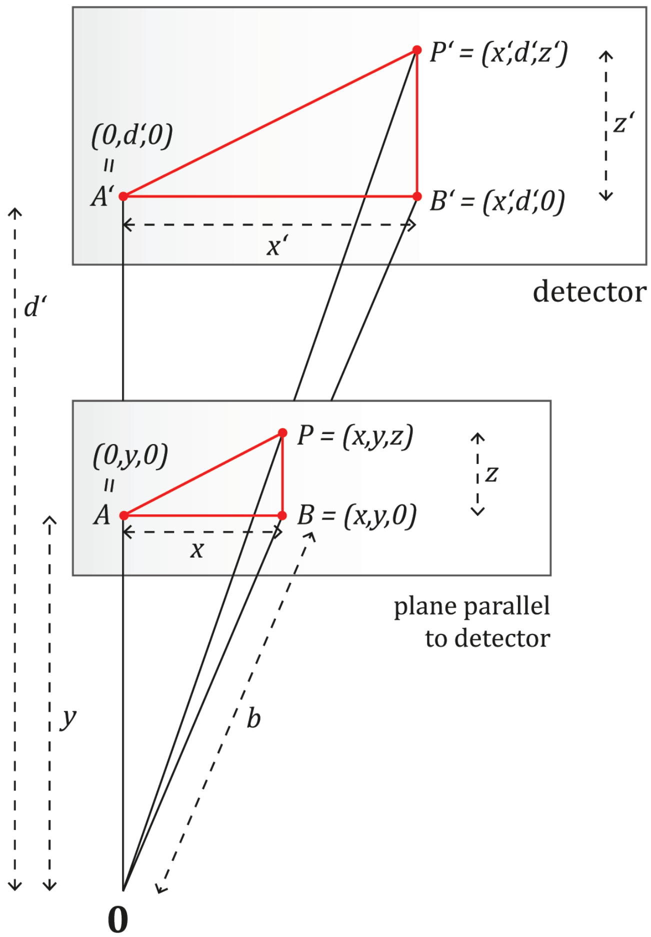

Fig. 2

Geometric drawing illustrating the intercept theorem used to derive the relationship between an arbitrary point $P = (x, y, z)$ and its image $P' = (x', d', z')$ located on the detector (first step of model construction). The Cartesian coordinate system has its origin O at the X-ray source and its y-axis coincides with the direction of the central X-ray beam. The source-image distance is denoted by d'.

placed in supine position. To achieve reproducible projections, a standardized protocol was followed in which both legs are rotated inwards by 15° to align the femoral neck parallel to the detector.

All images were obtained with the same radiograph tube (DigitalDiagnost 4.2; Philips Healthcare, the Netherlands). The source-image distance was set to 120 cm with the X-ray beam perpendicular to the table. First, the AP radiograph of the pelvis was obtained with the central X-ray beam centred on the patent's pubic symphysis. Second, the X-ray source was moved laterally and centred on the patient's femoral head. As a reference, the midpoint of a line between the anterior superior iliac spine (ASIS) and the pubic symphysis was used. The patient's position and the position of the legs remained unchanged during this procedure. All images were calibrated with the KingMark calibration system (Brainlab AG, Germany).[12] The calculated magnification was documented. All images were saved in a digital imaging and communications in medicine (DICOM) format on a picture archiving and communication system (PACS). The offset measurements were performed with a validated software (TraumaCAD; Brainlab AG).

On both AP hip and pelvis radiographs, the centre of the femoral head was determined. Two circles reaching the medial and lateral border of the femoral shaft were digitally drawn 20 mm below the lesser trochanter and at the level of the femoral isthmus. The connection of the two circle centres defined the femoral shaft axis. The perpendicular distance between the femoral shaft axis and the femoral head centre was defined as the femoral offset measured on an AP hip (FO_h) and AP pelvis (FO_p) radiograph (Figure 1).

Additionally, the mid-centre distance (MCD)[13] was measured on the AP pelvis radiograph, defined as the distance between the midline of the pelvis and the centre of the femoral head (Figure 1). Rotation of the leg was assessed on the AP pelvis radiograph using the thickness of the lesser trochanter (TLT) described by Hananouchi et al.[14] TLT was represented by the perpendicular distance from a line passing through the proximal and distal cortical intersection between the lesser trochanter and the femoral cortex, to a second line passing through the tip of the lesser trochanter (Figure 1). According to Hananouchi et al,[14] neutral to mild (≤ 15°), moderate, and severe (≥ 45°) external rotation of the femoral neck to the coronal plane were defined as TLT < 5 mm, TLT 5 to 10 mm, and TLT > 10 mm, respectively.

Geometric model. A geometric model was created to simulate the FOs depicted in AP hip and AP pelvis images, considering different leg positions. The step-by-step construction of the geometric model is explained below. A Cartesian coordinate system is defined with origin O at the X-ray source and y-axis coinciding with the direction of the central X-ray beam. The source-image distance is denoted by d'. The first step is to derive the geometrical relationship between an arbitrary point $P = (x, y, z)$ and its image $P' = (x', d', z')$ located on the detector. Figure 2 illustrates the intercept theorem used to solve this problem. The lines AB and A′B′ are parallel; the same is true for the lines BP and B′P′. Therefore, the intercept theorem states

$$\frac{x'}{x} = \frac{d'}{y}, \quad \frac{b'}{b} = \frac{d'}{y} \quad , and \quad \frac{z'}{z} = \frac{b'}{b}$$

or equivalently,

$$x' = \frac{d'}{y} \cdot x \quad and \quad z' = \frac{b'}{b} \cdot z = \frac{d'}{y} \cdot z$$

As a result, the image of $P = (x, y, z)$ reads

$$P' = \left(\frac{d'}{y} \cdot x, d', \frac{d'}{y} \cdot z\right) = \frac{d'}{y} \cdot (x, y, z) \qquad (1)$$

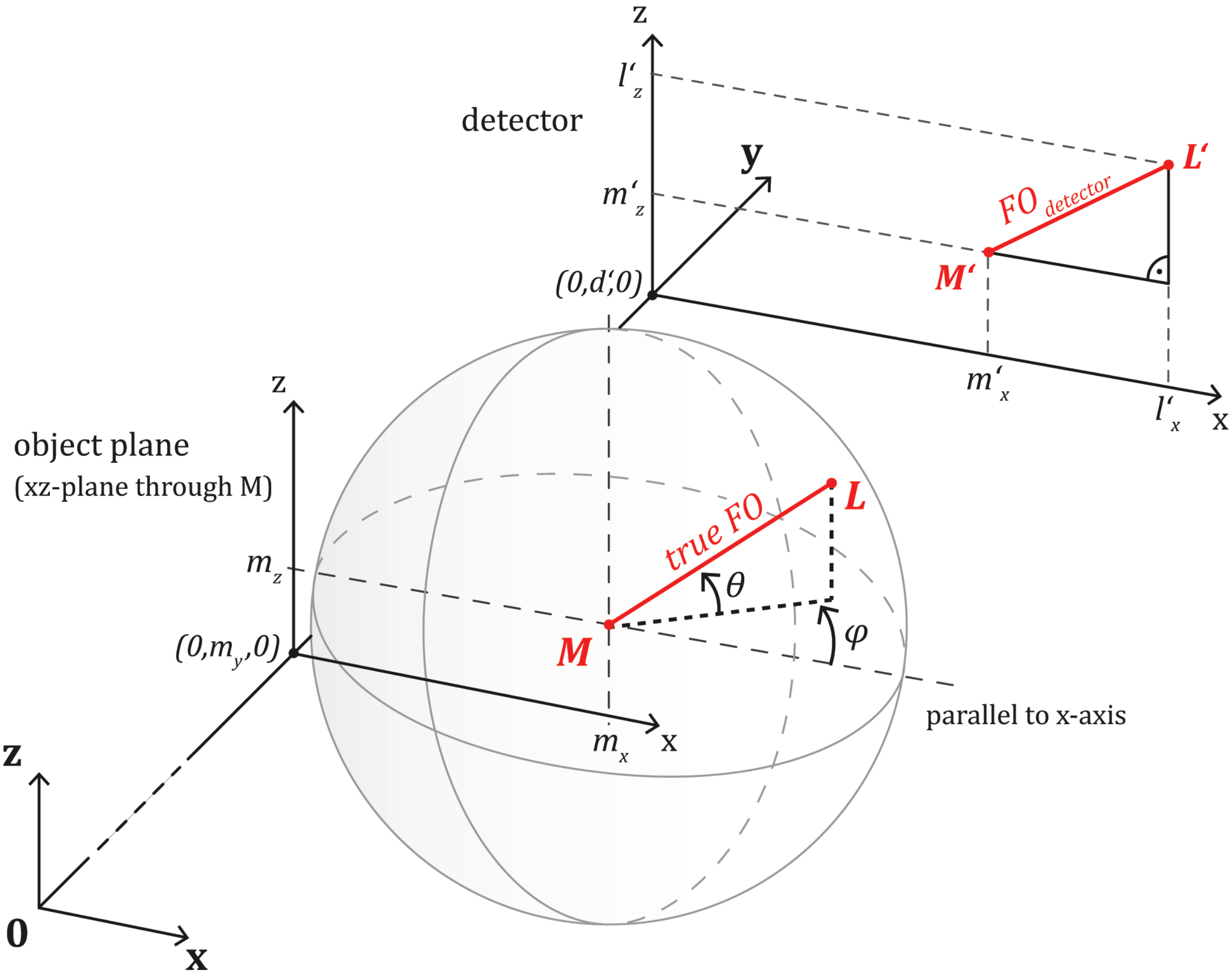

Fig. 3

Geometric drawing used for the second step of model construction. $M = (m_x, m_y, m_z)$ is the femoral head centre. $L = (l_x, l_y, l_z)$ lies on a sphere with centre M and radius equals true femoral offset (FO) and is determined by knowing two angles θ (leg abduction/adduction) and φ (femoral neck rotation). The FO projected onto the detector ($FO_{detector}$) is the distance between the points $M' = (m'_x, d', m'_z)$ and $L' = (l'_x, d', l'_z)$ and can be calculated using the Pythagorean theorem.

In a second step, this formula is used to calculate the FO projected onto the detector, taking into account different leg positions and rotations. We denote by $M = (m_x, m_y, m_z)$ the centre of the femoral head and by $L = (l_x, l_y, l_z)$ the point on the femoral shaft axis that minimizes the distance to M (Figure 3). By definition, the distance between M and L corresponds to the (true) femoral offset (FO_t). This means that L lies on a sphere with radius FO_t and centre M. Such a point is determined by knowing two angles θ (between -90° and +90°) and φ (between -180° and +180°) (Figure 3). In our model, θ was limited to -10° to +10° (ad- and abduction), and φ was limited to -15° to +45° (internal and external rotation of the femoral neck). Mathematically, the connection between L (given by θ and φ) and M is the following:

$$\begin{aligned} l_x &= m_x + FO_t \cdot \cos\theta \cdot \cos\varphi \\ l_y &= m_y + FO_t \cdot \cos\theta \cdot \sin\varphi \\ l_z &= m_z + FO_t \cdot \sin\theta \end{aligned} \qquad (2)$$

The distance on the detector between the points $M' = (m'_x, d', m'_z)$ and $L' = (l'_x, d', l'_z)$ can be calculated using the Pythagorean theorem (Figure 3) and is given by

$$FO_{detector} = \sqrt{(l'_x - m'_x)^2 + (l'_z - m'_z)^2}$$

Combining this with (1) and (2) yields

$$FO_{detector} = d' \cdot \sqrt{\left(\frac{m_x + FO_t \cdot \cos\theta \cdot \cos\varphi}{m_y + FO_t \cdot \cos\theta \cdot \sin\varphi} - \frac{m_x}{m_y}\right)^2 + \left(\frac{m_z + FO_t \cdot \sin\theta}{m_y + FO_t \cdot \cos\theta \cdot \sin\varphi} - \frac{m_z}{m_y}\right)^2}$$

In a final step, the FO is calculated in the object plane (xz-plane through M) by including the

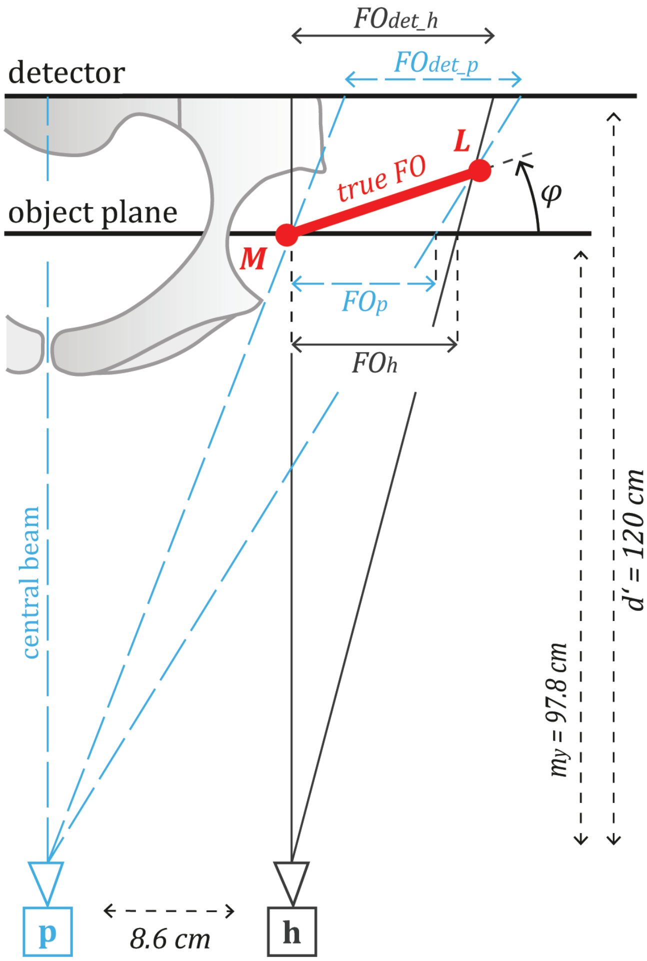

Fig. 4

Geometric drawing illustrating the two situations in which the central X-ray beam was directed to the femoral head (anteroposterior (AP) hip view) and to the pubic symphysis (AP pelvis view), respectively. For these two situations, the projected femoral offset (FO) onto the detector was labelled by FO_{det_h} and FO_{det_p} and the FO in the object plane was denoted by FO_h and FO_p, respectively. The absolute distances used for the simulation are given in cm. φ was limited to -15° internal to +45° external rotation of the femoral neck.

KingMark-calibration, i.e. divide by the magnification factor $MF = \frac{d'}{m_y}$

$$FO_{obj\_plane} = m_y \cdot \sqrt{\left(\frac{m_x+FO_t \cdot \cos\theta \cdot \cos\varphi}{m_y+FO_t \cdot \cos\theta \cdot \sin\varphi} - \frac{m_x}{m_y}\right)^2 + \left(\frac{m_z+FO_t \cdot \sin\theta}{m_y+FO_t \cdot \cos\theta \cdot \sin\varphi} - \frac{m_z}{m_y}\right)^2}$$

In the clinical setting, FO_{obj_plane} corresponds to the measured FO used for preoperative planning. For the special case where *M* and *L* both lie in the object plane (φ = 0°), the formula reduces to $FO_{obj\_plane} = FO_t$ no matter where the X-ray source is placed.

We finally applied our geometric model to the two situations in which the central X-ray beam was directed to the femoral head (AP hip view) and to the pubic symphysis (AP pelvis view), respectively. We denoted FO_{obj_plane} in these two situations by FO_h and FO_p, respectively (Figure 4). For the AP hip projection, we chose $m_x = m_z = 0$ by design. For the AP pelvis projection, we had $m_z = 0$, and m_x = 8.6 cm corresponding to the mean MCD measured in the current study (Figure 4). FO_t was assumed to be 40 mm or 50 mm. Moreover, the source-image distance was set to d' = 120 cm. KingMark-calibration revealed a mean magnification factor *MF* = 1.227, resulting in $m_y = \frac{d'}{MF} = 97.8\ cm$ (Figure 4). The magnification factor may also be calculated with a standard 25.5 mm (1 inch) marker ball. This was not performed within the scope of this study because the authors' institution uses King Mark calibration as standard. The simulations were carried out with MATLAB (MathWorks, USA).

Statistical analysis. To determine sample size, a power analysis was performed with G*Power for Mac (Version 3.1, University of Düsseldorf, Germany). Power analysis was performed with the following assumptions: normally distributed data, matched pairs, effect size 0.5, an α error of 0.05, and power of 0.95. The FO measurements were performed by two independent blinded observers (EFL, KK); both of them made the readings on two separate occasions at least two weeks apart. Intra- and interobserver reliabilities were calculated using single-measure intraclass correlation coefficients (ICCs) with a two-way random effects model for absolute agreement. Continuous variables are presented as means and standard deviation (SD). Conformity of data to normal distribution was evaluated with the Kolmogorov-Smirnov test. Group comparisons were performed using the paired *t*-test for paired observations and the independent-samples *t*-test for unpaired observations. One-way analysis of variance (ANOVA) was used for comparisons of more than two independent groups. Correlation of continuous variables was evaluated with the Pearson correlation coefficient. Significance was set at $p < 0.05$. Differences in FO measurements between AP hip and AP pelvis views were analyzed using a Bland-Altman plot. IBM SPSS Statistics v. 25 (IBM, USA) was used for statistical analysis.

Results

Radiological analysis. The intra- and interobserver ICC scores were excellent for FO_h and FO_p measured on AP hip and pelvis radiographs, respectively (Table I).

All collected radiological parameters are presented in Table II. The position of the X-ray source (AP hip view vs AP pelvis view) affected the FO measurements. The mean FO_h was 38.0 mm (SD 6.4) and the mean FO_p was 36.6 mm (SD 6.3) ($p < 0.001$, paired *t*-test). FO_h and FO_p were both significantly higher in men than in women (Table II). There was an excellent correlation between FO_h and FO_p (r = 0.980; $p < 0.001$, two-tailed *t*-distribution).

The differences between FO_h and FO_p were smallest in the group with TLT < 5 mm (neutral to mild external rotation of the femoral neck) and largest in the group with

Table I. Reliability and reproducibility of the femoral offset measurements.

Parameter (95% CI)	ICC intraobserver 1	ICC intraobserver 2	ICC interobserver
FO_h	0.946 (0.908 to 0.969)	0.920 (0.868 to 0.953)	0.953 (0.921 to 0.972)
FO_p	0.962 (0.930 to 0.979)	0.945 (0.908 to 0.968)	0.954 (0.923 to 0.973)

CI, confidence interval; FO_h, femoral offset on anteroposterior hip radiograph; FOp, femoral offset on anteroposterior pelvis radiograph; ICC, intraclass correlation coefficient.

Table II. Collected radiological data. The differences between femoral offset measured on anteroposterior hip and pelvis radiographs are grouped by thickness of the lesser trochanter.

Parameter	n (%)	Mean (95% CI)	SD	Minimum	Maximum	p-value
FO_h, mm						
Overall	55 (100)	38.0 (36.3 to 39.7)*	6.4	21.5	54.3	< 0.001*
Males	31 (56)	39.9 (38.4 to 41.5)†	5.7	30.1	54.3	0.010†
Females	24 (44)	35.5 (33.8 to 37.2)†	6.6	21.5	48.9	
FO_p, mm						
Overall	55 (100)	36.6 (34.9 to 38.3)*	6.3	20.7	53.3	
Males	31 (56)	38.2 (36.7 to 39.7)†	5.7	26.3	53.3	0.034†
Females	8 (44)	34.6 (32.8 to 36.3)†	6.6	20.7	48.8	
$FO_h - FO_p$, mm						
Overall	55 (100)	1.38 (1.04 to 1.72)	1.30	-1.44	4.31	
TLT < 5	12 (22)	0.52 (0.23 to 0.82)‡	1.10	-1.44	1.98	< 0.001‡
TLT 5 to 10	35 (64)	1.40 (1.09 to 1.71)‡	1.18	-0.57	3.92	
TLT > 10	8 (14)	2.58 (2.25 to 2.90)‡	1.22	0.73	4.31	
TLT, mm		7.09 (6.29 to 7.90)	3.04	-2.79	14.23	
MCD, mm		85.9 (83.8 to 87.9)	7.8	65.0	103.0	
MF		1.23 (1.22 to 1.23)	0.02	1.18	1.29	

*Significant difference between FO_h and FO_p (paired samples *t*-test).
†Significant differences between males and females (independent-samples *t*-test).
‡Significant differences within the three TLT groups (one-way analysis of variance).
CI, confidence interval; FO_h, femoral offset on AP hip radiograph; FO_p, femoral offset on AP pelvis radiograph; MCD, mid-centre distance; MF, magnification factor; SD, standard deviation; TLT, thickness of the lesser trochanter.

severe external neck rotation (TLT > 10 mm; p < 0.001, one-way ANOVA) (Table II). When comparing AP hip and AP pelvis radiographs, FO measurements were found to agree within ± 2 mm in 70.9% of cases (39/55) (Figure 5). Exclusion of cases with TLT > 10 mm increased agreement within ± 2 mm to 78.7% (37/47). Considering only the cases with TLT < 5 mm resulted in an agreement of 100% (12/12) (Figure 5).

Geometric model. Figure 6 shows the values of FO_h (grey) and FO_p (blue) in relation to leg position (θ and φ). FO_h and FO_p were strongly influenced by φ (femoral neck rotation), whereas θ (ad- and abduction) had negligible influence on FO_h and FO_p. In the setting of neutral femoral neck alignment parallel to the detector ($\varphi = 0°$), FO_h and FO_p were equal and represented the true FO (plane-plane intersection in Figure 6).

Figure 7 shows the values of FO_h and FO_p as a function of φ (the value of θ was set to 0°) for the two different true FOs of 40 and 50 mm. There was a non-linear relationship between φ and projected FO, viz. the larger φ became, the more FO_h and FO_p underestimated the true FO. In the case of $\varphi = 30°$ (femoral neck external rotation with reference to the detector plane), FO_h and FO_p underrated a true FO of 40 mm by 15% (6.1 mm) and 20% (7.8 mm), respectively. At $\varphi = 45°$ (severe external neck rotation), FO_h and FO_p underestimated a true FO of 40 mm by 31% (12.5 mm) and 37% (14.9 mm), respectively. With a neutral to mild neck external rotation ($\varphi \leq 15°$) the underestimation was less than 7% (2.7 mm).

The position of the X-ray source (AP hip view vs AP pelvis view) also affected the FO measurements ($\varphi \neq 0°$). However, the difference between AP hip and AP pelvis views ($FO_h - FO_p$) was small, being only 0.9 mm, 1.7 mm, and 2.4 mm for φ equal to 15°, 30°, and 45°, respectively (Figure 8). Furthermore, the influence of low-centred AP radiographs on the projected FO was negligible (results for simulations with m_z = 7.5 cm and m_z = 15 cm are provided in Supplementary Figure a).

Discussion

Accurate restoration of the physiological biomechanics in THA is a key factor for good functional outcome and favours the longevity of the implants.[7,8] Accordingly, the accurate determination of the FO is of eminent importance in the preoperative planning of THA. AP radiographs of the pelvis are the accepted method for planning

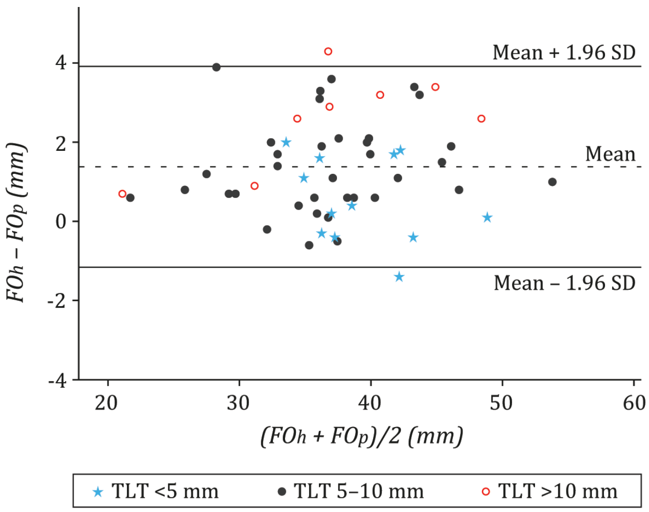

Fig. 5

Bland-Altman plot illustrating the agreement between femoral offset (FO) measurements on anteroposterior (AP) hip (FO_h) and AP pelvis (FO_p) radiographs. The cohort was grouped by the thickness of the lesser trochanter (TLT). SD, standard deviation.

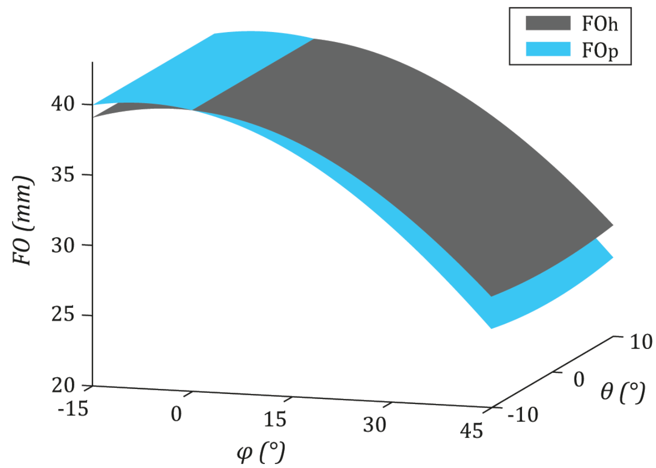

Fig. 6

Theoretical femoral offset (FO) on anteroposterior (AP) hip (FO_h, grey) and AP pelvis (FO_p, blue) radiographs as a function of leg abduction/adduction (θ) and femoral neck rotation (φ). The true FO was assumed to be 40 mm. FO_h and FO_p were strongly influenced by φ, whereas θ influenced them only slightly. In the setting of $\varphi = 0°$, FO_h and FO_p were equal and represented the true FO (plane-plane intersection).

THA because they provide important information about both hip joints, and are critical for restoring leg length and centre of rotation. However, Merle et al[10] demonstrated that projection errors may occur depending on the X-ray source position and therefore recommended additional routine AP hip radiographs. Furthermore, it is well established that leg rotation during X-ray imaging affects the projected FO.[15-17] To date, the relationship between leg rotation and projected FO has been investigated only for AP hip views using a simplified mathematical model.[17] To our knowledge, this is the first study presenting quantitative data of the measurement error of the FO. To this end, a geometric model evaluating both the effects of leg position (rotation and abduction/adduction) and the effects of X-ray source position on FO measurement was developed.

According to our model, inaccurate patient positioning significantly affects FO measurements in both AP hip and pelvis views; at 30° external rotation of the femoral neck (with reference to the detector plane, $\varphi = 30°$), FO is underestimated in the order of 15% and 20%, respectively. With severe neck external rotation ($\varphi = 45°$), the measurement error is even almost twice as large. This non-linear relationship between projected FO and femoral neck rotation is consistent with the calculations of Lechler et al.[17] It should be noted that the higher the true FO is (e.g. large body size, coxa vara), the larger the absolute measurement error (underestimation) will be. Other than external rotation, abduction and adduction of the leg during image acquisition has no clinical significance for FO measurement.

The position of the X-ray source (AP hip view or AP pelvis view) has only a minor influence on the measurement of the FO. On average, FO_p was only 1.38 mm smaller than FO_h in our patient cohort. In particular, with neutral to mild external rotation of the femoral neck ($\varphi \leq 15°$), the difference is likely to be of little clinical importance. When the femoral neck is aligned parallel to the detector, the true FO is measured independently of the radiological view.

Typically, hip implants have a FO difference of 3 mm between sizes. In the case of moderate external rotation of the femoral neck ($\varphi = 30°$) during pelvis imaging, this would result in a planning error of about three implant sizes. Severe external rotation of the femoral neck ($\varphi = 45°$) would lead to a deviation of five to six implant sizes. From the authors' point of view, a planning error of ± 1 implant size is acceptable. According to our geometric model, this can be achieved by limiting the malrotation of the femoral neck (φ) to less than 15°, regardless of the radiological view.

Based on these considerations, an AP pelvis view is suitable for routine THA templating as long as the femoral neck is aligned to the detector in the range of ± 15°. To reduce the effect of femoral antetorsion during image acquisition, the leg must be rotated internally by 15° to 20°.[14] Although standardized protocols with defined internal rotation of the leg are crucial, not every patient will have a correctly aligned femoral neck. Common causes include high femoral ante- or retrotorsion, as well as unpredictable compensatory effects of tibial torsion when using the foot progression angle for

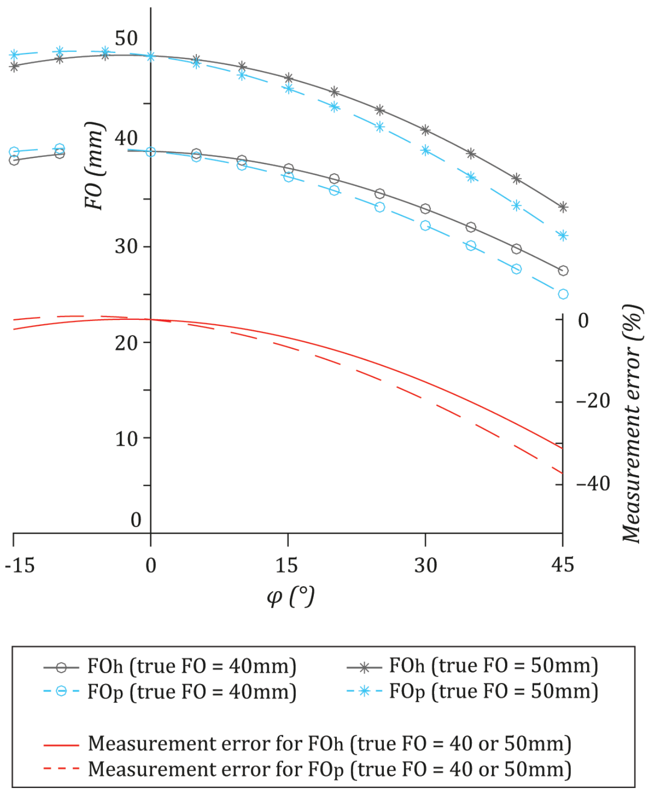

Fig. 7

Theoretical femoral offset (FO) on anteroposterior (AP) hip (FO_h, grey) and AP pelvis (FO_p, blue) radiographs as a function of femoral neck rotation (φ). The simulation was run twice for a true FO of 40 mm and 50 mm, respectively. The relative measurement error in % (y-axis on the right) was calculated for both simulations and found to be almost identical (only one curve is shown for clarity). There was a non-linear relationship between φ and projected FO. The larger φ was, the more the true FO was underestimated (represented by a negative measurement error).

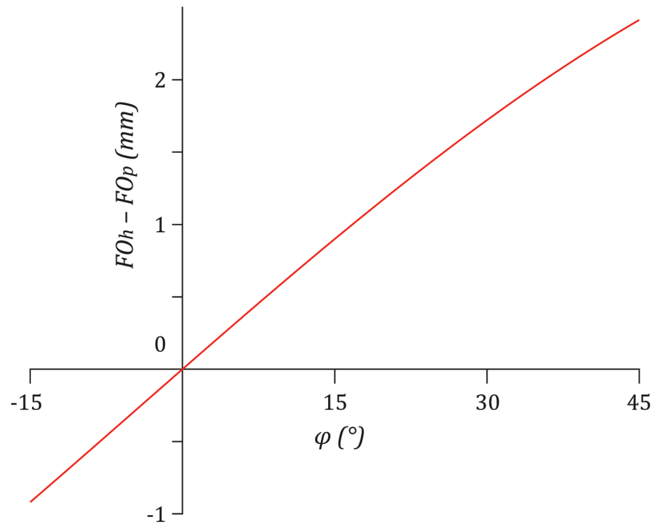

Fig. 8

Difference between the theoretical femoral offsets (FO) for two different X-ray source positions (anteroposterior (AP) hip view vs AP pelvis view). The true FO was assumed to be 40 mm. The difference $FO_h - FO_p$ was small, with a maximum value of 2.4 mm for severe external rotation of the femoral neck ($\varphi = 45°$).

leg positioning. For this group of patients, we recommend repeating the AP pelvis view with appropriate adjustment of leg rotation. Other reasons for inaccurate leg positioning are external rotation contracture and pain due to end-stage OA of the hip. As pointed out by Merle et al,[10] these problems can be addressed by an additional AP hip radiograph with elevation of the entire affected hip (wedge under the buttock) to align the femoral neck more parallel to the detector. Alternatively, CT scans may be performed to measure FO preoperatively in complicated cases.[9,18]

Measurement of the thickness of the lesser trochanter has been shown to be a good tool for assessing the extent of leg rotation.[14] Our subgroup with a TLT < 5 mm showed a mean difference between FO_h and FO_p of 0.52 mm (Table II), which was consistent with the geometric model for neutral to mild externally rotated femoral neck ($\varphi \leq 15°$). The same was true for the two other subgroups representing moderate and severe external neck rotation, respectively. This is in accordance with the findings of Hananouchi et al.[14] Another tool, called the 'lesser trochanter index', has been proposed for predicting underestimation of FO in AP radiographs of the pelvis.[19] However, we were not able to reliably assess this index from radiographs, which is in line with the findings of another research group.[20] One possible reason may be that the index was developed using simulated radiographs from volumetric CT data, facilitating the definition of the anatomical landmarks.[19]

We acknowledge the following limitations of the study. The radiological analysis was conducted on a consecutive case series with its inherent limitations. Despite a standardized protocol for obtaining radiographs, both the positioning of the leg and the positioning of the X-ray source are dependent on the technician's judgment and thus represent a potential source of bias. Furthermore, pain, contractures, or limited patient compliance during radiography were not recorded. The patient's individual femoral torsion was not investigated, which would have required an additional rotational CT scan or MRI. Lastly, the radiological analysis could only reveal the influence of the X-ray source position ($FO_h - FO_p$),whereas the absolute measurement error could not be calculated because the patient's true FO was unknown. However, we do not consider this a major limitation, as our geometric model answered this question.

In conclusion, the AP pelvis view remains the gold standard for routine THA templating. Only patients with femoral neck malrotation > 15° on the AP pelvis view, e.g. due to external rotation contracture, should receive further imaging. A good tool for decision-making in this regard is the TLT, and values of ≥ 5 mm should be evaluated thoroughly. In such cases, an additional AP hip view with elevation of the affected hip to align the femoral neck more parallel to the detector is a simple option. Whether this measure is enough or CT is needed in cases with very severe femoral neck malrotation should be subject of future investigations.

Take home message

- Anteroposterior (AP) pelvis radiography remains the gold standard for routine total hip arthroplasty templating.
- Measurement of the thickness of the lesser trochanter is a good tool for assessing the extent of femoral neck rotation.
- Femoral neck malrotation > 15° on the AP pelvis view should prompt further imaging.

Supplementary material

Theoretical influence of low-centred X-ray source on projected femoral offset on anteroposterior hip and pelvis radiographs.

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Author information:
- E. F. Liechti, MD, MSc, Orthopaedic Surgeon
- M. C. Attinger, MD, Orthopaedic Surgeon
- A. Hecker, MD, Orthopaedic Surgeon
- K. Kuonen, MD, Research Assistant
- A. Michel, MSc, Research Assistant
- F. M. Klenke, MD, PhD, Orthopaedic Surgeon
  Department of Orthopaedic Surgery and Traumatology, Inselspital, Bern University Hospital, University of Bern, Bern, Switzerland.

Author contributions:
- E. F. Liechti: Conceptualization, Methodology, Investigation, Data curation, Formal analysis, Visualization, Writing – original draft, Writing – review & editing.
- M. C. Attinger: Conceptualization, Methodology, Writing – original draft.
- A. Hecker: Writing – review & editing.
- K. Kuonen: Investigation.
- A. Michel: Methodology, Formal analysis.
- F. M. Klenke: Conceptualization, Methodology, Resources, Supervision, Writing – review & editing.

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